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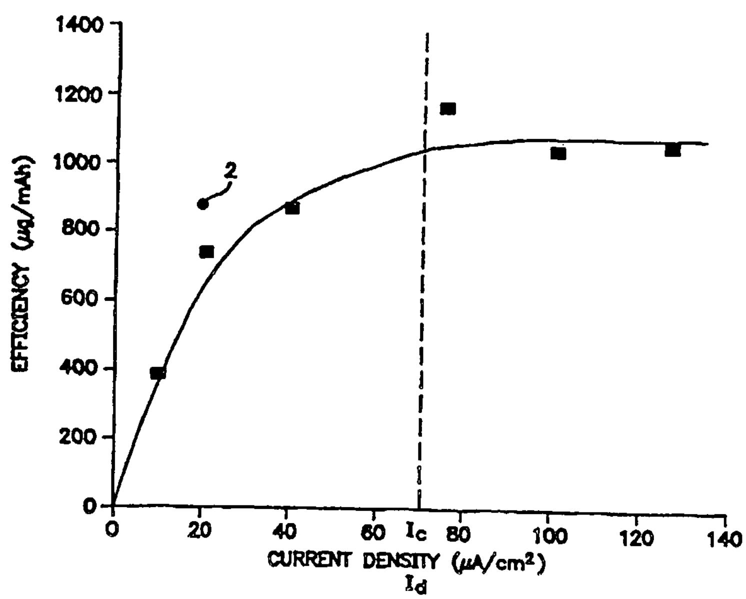

(57) Abstract

An electrotransport agent delivery device (10) for delivering a therapeutic agent through intact skin, and a method of operating same, is provided. The device applies a pulsing electrotransport current wherein the length of the applied current pulses is at least 5 msec and preferably at least 10 msec. Most preferably, the current pulses have a magnitude above a critical level (I_c) at which the skin is transformed into a higher electrotransport delivery efficiency (E) state.

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ELECTROTRANSPORT AGENT DELIVERY METHOD AND APPARATUS

TECHNICAL FIELD

The present invention generally concerns a method and apparatus for the electrically assisted delivery of a therapeutic agent (e.g., a drug) through a body surface (e.g., skin) at increased efficiency. This invention is particularly applicable to the electrotransport of highly potent therapeutic agents which are to be delivered at small dosage levels.

BACKGROUND OF THE INVENTION

The present invention concerns in vivo methods and apparatuses for electrotransport delivery of therapeutic agents, typically drugs, into a patient. Herein the terms "electrotransport", "iontophoresis" and "iontophoretic" are used to refer to methods and apparatus for transdermal delivery of therapeutic agents, whether charged or uncharged, by means of an applied electromotive force to an agent-containing reservoir. The particular therapeutic agent to be delivered may be completely charged (i.e., 100% ionized), completely uncharged, or partly charged and partly neutral. The therapeutic agent or species may be delivered by electromigration, electroosmosis or a combination of these processes. Electroosmosis has also been referred to as electrohydrokinesis, electro-convection, and electrically-induced osmosis. In general, electroosmosis of a therapeutic species into a tissue results from the migration of solvent, in which the species is contained, as a result of the application of electromotive force to a reservoir containing the therapeutic species.

As used herein, the terms "electrotransport", "iontophoresis" and "iontophoretic" refer to (1) the delivery of charged drugs or agents by electromigration, (2) the delivery of uncharged drugs or agents by the

process of electroosmosis, (3) the delivery of species by transport processes which include an electroporation step (See, e.g., Weaver et al US Patent 5,019,034), (4) the delivery of charged drugs or agents by the combined processes of electromigration and electroosmosis, and/or (5) the delivery of a mixture of charged and uncharged drugs or agents by the combined processes of electromigration and electroosmosis, combinations of the above processes to deliver either or both of charged or uncharged species.

Iontophoretic devices for delivering ionized drugs through the skin have been known since the early 1900's. See for example Deutsch US Patent 410,009. In presently known electrotransport devices, at least two electrodes or electrode assemblies are used. Both electrodes/electrode assemblies are disposed so as to be in intimate electrical contact with some portion of the skin of the body. One electrode, called the active or donor electrode, is the electrode from which the ionic substance, agent, medicament, drug precursor or drug is delivered into the body through the skin by iontophoresis. The other electrode, called the counter or return electrode, serves to close the electrical circuit through the body. In conjunction with the patient's skin contacted by the electrodes, the circuit is completed by connection of the electrodes to a source of electrical energy, e.g., a battery. For example, if the ionic substance to be delivered into the body is positively charged, then the positive electrode (the anode) will be the active electrode and the negative electrode (the cathode) will serve to complete the circuit. If the ionic substance to be delivered is negatively charged, then the cathodic electrode will be the active electrode and the anodic electrode will be the counter electrode.

As is discussed above, electrotransport delivery devices can be used to deliver uncharged drugs or agents into the body, e.g., transdermally. This is accomplished by a process called electroosmosis. Electroosmosis is the (e.g., transdermal) flux of a liquid solvent (e.g., the liquid solvent containing

the uncharged drug or agent) which is induced by the presence of an electric field imposed across the skin by the donor electrode.

Electrotransport electrode assemblies/devices generally include a reservoir or source of the beneficial agent or drug (preferably an ionized or ionizable species or a precursor of such species), which is to be delivered into the body by electrotransport. Examples of such reservoirs or sources include a pouch as described in Jacobsen US Patent 4,250,878, a pre-formed gel body as disclosed in Webster US Patent 4,383,529 and Ariura, et al US Patent 4,474,570 and a receptacle containing a liquid solution as disclosed in Sanderson, et al US Patent 4,722,726. Such drug reservoirs are connected to the anode or the cathode of an electrotransport device to provide a fixed or renewable source of one or more desired species or agents. Electrical current is typically applied to the reservoir by means of a current distributing member, which may take the form of a metal plate, a foil layer, a conductive screen, or a polymer film loaded with an electrically conductive filler such as silver or carbon particles. The current distributing member, including any appropriate connectors and associated connective conductors such as leads, and the reservoir comprise an electrode assembly herein.

The prior art has recognized that "competitive" ionic species having the same charge (i.e., the same sign) as the drug ions being delivered by electrotransport have a negative impact on electrotransport drug delivery efficiency. The efficiency (E) of electrotransport delivery of a particular species is defined herein as the rate of electrotransport delivery of that species per unit of applied electrotransport current (mg/mA-h). The prior art further recognized that competitive ionic species were inherently produced during operation of these devices. The competitive species produced are dependent upon the type of electrode material which is in contact with the drug solution. For example, if the electrode is composed of an electrochemically inert material (e.g., platinum or stainless steel), the

electrochemical charge transfer reaction occurring at the electrode surface tended to be water electrolysis since water is the overwhelmingly preferred liquid solvent used in electrotransport drug solutions. Water electrolysis produces competing hydronium ions at the anode (in the case of cationic electrotransport drug delivery) and competing hydroxyl ions at the cathode (in the case of anionic electrotransport drug delivery). On the other hand, if the electrode is composed of an electrochemically oxidizable or reducible species, then the electrode itself is oxidized or reduced to form a competitive ionic species. For example, Untereker et al US Patent 5,135,477 and Petelenz et al US Patent 4,752,285 recognize that competitive ionic species are electrochemically generated at both the anode and cathode of an electrotransport delivery device. In the case of an electrotransport delivery device having a silver anodic donor electrode, application of current through the silver anode causes the silver to become oxidized ($Ag \rightarrow Ag^+ + e^-$) thereby forming silver cations which compete with the cationic drug for delivery into the skin by electrotransport. The Untereker and Petelenz patents teach that providing a cationic drug in the form of a halide salt causes a chemical reaction which removes the "competing" silver ions from the donor solution (i.e., by reacting the silver ions with the halide counter ion of the drug to form a water insoluble silver halide precipitate; $Ag^+ + X^- \rightarrow AgX$), thereby achieving higher drug delivery efficiency. In addition to these patents, Phipps et al PCT/US95/04497 filed on April 7, 1995 teaches the use of supplementary chloride ion sources in the form of high molecular weight chloride resins in the donor reservoir of a transdermal electrotransport delivery device. These resins are highly effective at providing sufficient chloride for preventing silver ion migration, yet because of the high molecular weight of the resin cation, the resin cation is effectively immobile and hence cannot compete with the drug cation for delivery into the body.

The prior art has long recognized that the application of electric current through skin causes the electrical resistance of the skin to decrease. See, for example, Haak et al US Patent 5,374,242 (Figure 3). Thus, as the electrical resistance of the skin drops, lower voltages are needed to drive a particular level of electrotransport current through the skin. This same phenomenon is observed in a technique referred to as "electroporation" of the skin. See Weaver et al US Patent 5,019,034. Electroporation involves the application of short, high voltage electrical pulses to produce what is characterized as a transient (e.g., decreasing to normal levels in 10 to 120 sec. for excised frog skin) increase in tissue permeability. Electroporation is also characterized by the creation of pores in lipid membranes due to reversible electrical breakdown. Electroporation does not, itself, deliver any drug but merely prepares the tissue thereby treated for delivery of drug by any of a number of techniques, one of which is iontophoresis.

DISCLOSURE OF THE INVENTION

The present invention arises from the discovery that when delivering a therapeutic agent (eg, a drug) via electrotransport through a living body surface (eg, skin) of an animal (eg, a human) using a pulsing electrotransport current, the efficiency of electrotransport agent delivery is increased by maintaining the width of the applied current pulses above a minimum period of time. For certain drugs delivered transdermally to humans via electrotransport, this minimum period has been found to be about 5 msec, and preferably about 10 msec. In general, this discovery means that lower frequency pulsing electrotransport currents tend to provide more efficient agent delivery than higher frequency pulsing electrotransport currents, since the longer the pulse width, the fewer the number of pulses which can be applied in any unit of time. Thus, when using pulsing currents having pulse

widths of at least about 5 msec, and preferably at least about 10 msec, the pulsing frequencies tend to be less than about 100 Hz and more preferably less than about 10 Hz.

As used herein, the term "electrotransport agent delivery efficiency (E)" means the rate of transdermal electrotransport delivery (mg/h) per unit of applied electrotransport current (mA) and expressed in units of micrograms of agent (i.e., drug) delivered per milliamp-hour of applied electric current (µg/mAh). Electrotransport delivery efficiency, in some aspects of its meaning, is analogous to transport number. Transport number is a unitless quantity, less than one, indicating the fractional charge carried by a particular ionic species, e.g., a drug or agent, during electrotransport delivery. Electrotransport delivery efficiency, as defined herein, is more broadly applicable to include the transport of uncharged species and is more reflective of the scope of the invention.

The terms "pulsing current" and "pulsed current" as used herein refer to an applied electrotransport current having a periodic (i.e., the waveform repeats over time and has a wave length and a frequency) waveform shape comprised of a first segment of applied electrotransport current having a first average current magnitude, and a second segment of applied electrotransport current having a second average current magnitude, the second average current magnitude being less than the first average current magnitude. In general, the second average current magnitude is less than about 70% of the first average current magnitude, more typically less than about 50% of the first average current magnitude and most typically less than about 25% of the first average current magnitude. The second average current magnitude can be zero or substantially zero, but in any event is substantially less than the first average current magnitude.

The present invention is not limited to any particular periodic pulsed waveform shape and may take the form of any of various types of periodic waveforms including sinusoidal, trapezoidal, square or rectangular current

waveforms. A square pulsed current waveform shape is particularly suitable for practicing this invention.

In a preferred embodiment of the present invention, the first average current magnitude is sufficient to produce a current density which is equal to or greater than a critical current density, I_c. Applied electrotransport current densities (generally expressed in units of microamperes per square centimeter ($\mu A/cm^2$) herein) above this critcal level result in even further enhancement of electrotransport transdermal agent delivery efficiency. This "further" enhancement of the skin's electrotransport delivery efficiency has been found to be non-transitory, i.e., to last for at least several minutes to several hours or longer after application of current densities and over periods of time in accordance with this preferred embodiment of the invention. This preferred embodiment of the invention induces (e.g., through a pre-treatment or pre-application step in which species are delivered) a high efficiency drug-transmissive state in the skin to which an electrotransport drug delivery device is applied. The induced, high efficiency state continues and can be utilized to deliver drug or other therapeutic agent transdermally with increased efficiency. In usual circumstances, this will permit delivery of drug with more precise control and at a lower current. This phenomenon has only been found in the transdermal delivery of drug or agent through intact living skin or tissue (i.e., in vivo) and is not exhibited in dead skin (i.e., excised skin through which species are electrotransported in vitro). In this manner, the treated skin exhibits a statistically significant, non-transitory increase in drug delivery efficiency relative to skin which has not been so treated. Generally speaking, utilization of this preferred embodiment of the invention significantly increases the drug/agent delivery efficiency and reduces or eliminates variability in the drug delivery efficiency of the skin site which is so treated. Since electrotransport delivery efficiency remains elevated and less variable after utilization of this embodiment (relative to untreated skin), utilization of this

embodiment of the invention permits the delivery of drug or agent through intact skin by electrotransport with increased control and efficiency.

Thus, in one aspect, the present invention is a method of electrotransport drug or agent delivery through a body surface involving the steps of delivering a therapeutic agent by a pulsing electrotransport current, the current pulses being sufficiently long (i.e., at least about 5 msec and preferably at least about 10 msec), to reduce or avoid capacitive loss and thereby deliver the agent at an enhanced electrotransport delivery efficiency (E). In a preferred aspect, the current pulses have a sufficient magnitude to produce a current density greater than or equal to I_c, to convert the electotransport delivery efficiency of the body surface (i.e., the skin) through which the agent is delivered to a non-transitory state of higher electrotransport delivery efficiency. Thereafter, the drug or agent is delivered through the body surface while the body surface is in the higher efficiency transfer state.

BRIEF DESCRIPTION OF THE DRAWINGS

A better understanding of the present invention, as well as other objects and advantages thereof, will become apparent upon consideration of the following modes for carrying out the invention especially when taken with the accompanying drawings, wherein:

FIG. 1 is a graph of transdermal electrotransport drug delivery efficiency (E) versus applied electrotransport current density (I_d) for in vivo electrotransport transdermal delivery of fentanyl;

FIG. 2 is a graph of electrotransport current versus time, showing three pulsed current waveforms having the same pulsing frequency but differing pulse widths and duty cycles;

FIG. 3 is an exploded perspective view of a transdermal electrotransport drug delivery device which can be used in accordance with the method of the present invention;

FIG. 4 is a graph of electrotransport current versus time, showing two pulsed waveforms having the same peak current and pulse width but different pulsing frequencies;

FIG. 5 is a graph of mean serum fentanyl concentration versus time, showing how initial electrotransport administered doses increase subsequent fentanyl delivery through a 24 hour period;

FIG. 6 is average serum fentanyl concentration, as a function of time, for applied electrotransport current densities of 10, 20 and 40 $\mu A/cm^2$;

FIG. 7 is a graph of serum fentanyl concentration versus time for delivery of fentanyl at pulsing frequencies of 1, 10 and 625 Hz; and

FIG. 8 is a graph of serum goserelin concentration versus time, for applied electrotransport current densities of 50 and 100 $\mu A/cm^2$.

MODES FOR CARRYING OUT THE INVENTION

The present invention is based upon the discovery that when delivering an agent (e.g., a drug) transdermally through intact skin via electrotransport using a pulsing electrotransport current, the efficiency (E) of transdermal electrotransport agent (e.g., drug) delivery is increased by maintaining the width of the current pulses greater than 5 msec and preferably greater than 10 msec. Since pulse width is inherently related to pulsing frequency, the discovery means that the efficiency of electrotransport delivery, when using a pulsing current, is greater at lower pulsing frequencies. Preferably, the pulsing frequency is maintained below about 100 Hz, and more preferably less than about 10 Hz. By maintaining longer pulse widths (and correspondingly lower pulsing frequencies), the inefficiencies associated with "charging up" the electrical capacitance of the skin are minimized. These

inefficiencies, termed "capacitive loss", are described in McNichols et al US Patent 5,047,007. Capacitive loss results because a portion of each pulse is consumed by the process of charging the skin without delivering drug. The shorter the pulse width (and hence the higher the pulsing frequency), the relatively greater is the capacitive loss for each pulse.

In a preferred practice, the electrotransport current density during the first segment and the length of the first segment are selected to maintain the higher efficiency species delivery state of the body surface (e.g., skin). This invention also includes the preferred practice of intentionally renewing the highly efficient species delivery state so as to optimize drug delivery efficiency if drug or agent delivery conditions are used which do not periodically renew it. In another preferred practice, the present invention is utilized to deliver drug or agent transdermally, i.e., through intact skin. In yet a further preferred practice, the present invention is used to deliver drug or agent through intact, live, human skin.

In this preferred practice of this invention, the precise current density and treatment time period needed to convert untreated skin to a highly transmissive state have been found to be fairly specific to the drug or therapeutic agent to be delivered. However, for the electrotransport delivery of analgesics using a pulsing electrotransport current, a pulse width of at least 10 msec at a current density of about 40 $\mu A/cm^2$, preferably at least about 50 $\mu A/cm^2$ and most preferably at least about 70 $\mu A/cm^2$ appears to convert the body site so treated to a highly efficient drug transmissive state. This preferred embodiment of the invention arises out of the discovery that electrotransport delivery efficiency is highly dependent (i.e., it is non-constant) at current densities in the range of about 0 to about 30 $\mu A/cm^2$, is moderately dependent upon current density in the range of about 40 to about 70 $\mu A/cm^2$ and is relatively independent of current density at current densities in excess of about 70 $\mu A/cm^2$. This unexpected change in efficiency (in theory, efficiency is not predicted to change with increasing current density) permits

electrotransport transdermal delivery of drug with significantly enhanced electrotransport delivery efficiency.

A second unexpected result is achieved in this preferred practice of the present invention, i.e., the change of the skin to the higher efficiency transmissive state is non-transitory with the skin remaining in the higher, and more stable, efficiency state for minutes to hours after the initial transformation, even in cases where the subsequently applied electrotransport current density is lowered to a level below I_c or turned off, completely. In other words, when the skin site has been converted to a highly efficient agent transmissive state by applying a pulsing electric current over pulse widths of at least 5 msec, and at or above current density I_c, reduction in applied electrotransport current (and therefore current density) does not cause the skin to immediately return to its initial, lower electrotransport delivery efficiency state. This observation respecting in vivo drug delivery is critically important to electrotransport system design.

The term "non-transitory" as used herein, when referring to the high efficiency electrotransport agent delivery state, means of sufficient length to permit drug to be delivered to achieve a therapeutic effect, generally at least several minutes and preferably at least an hour. Thus, for example, a relatively inexpensive ionic species may be used to trigger conversion of, e.g., a skin site, to a highly efficient and more stable ionic species delivery state, and thereafter relatively more expensive drug or agent may be delivered at greater efficiency and stability by electrotransport. Where the drug or agent is inexpensive, it may be used to convert the body delivery site to the highly efficient and more stable state, and thereafter may be delivered with greater efficiency, i.e., at lower current density and at greater stability.

The term "high/higher efficiency state" as used herein means conversion of any particular body or skin site to a state in which drug or agent delivery is at least 10% (preferably at least 20%) more efficient than the same skin site prior to conversion in accordance with this invention. Generally, the

parameter which will be most reflective of this efficiency increase will be the electrotransport delivery efficiency measured in micrograms of drug delivered per milliamp-hour of applied electrotransport current.

The term "more stable efficiency" as used herein means conversion from a state of more variable electrotransport agent delivery efficiency to one of less variability by exposure of the body site to a current density above the critical current density , I_c, for a time period longer than the critical time, t_c. Critical current density for purposes of increased stability, has been found to be as low as about 40 $\mu A/cm^2$.

The transdermal drug flux achieved by delivering drug at higher electrotransport delivery efficiency (i.e., at electrotransport current densities above the critical level I_c) may in some cases be higher than the flux needed to achieve the desired therapeutic effect. In such cases, it is desirable to reduce the transdermal drug flux, without reducing the electrotransport current density below the critical level I_c, so as to maintain the skin in the high efficiency and high stability transfer state. This problem may be overcome by one or more of the following three methods.

The first method of reducing the drug flux without reducing the applied level of electrotransport current, and hence current density, is to deliver the drug using a pulsing electrotransport current, the pulses of current producing a current density above I_c. and adjusting the pulse width of the current pulses (i.e., adjusting the duty cycle) in order to control the drug delivery rate. The term "duty cycle" as used herein is the ratio of the first period length (in msec) to the sum of the lengths of the first and second periods and is usually expressed as a percent. In other words, the duty cycle is the ratio of pulse width to cycle length. For example, if a device applies current pulses of 500 msec duration at a pulsing frequency of 1 pulse per second (i.e., 1 Hz), then the device is operating in a 50% duty cycle. In general, pulsing electrotransport currents typically have duty cycles of 10 to 95%, more typically 20 to 90%, and most typically 30 to 90%. In this practice of the

invention, the magnitude of the current pulses is selected in view of the known area of the surface from which drug is delivered, thereby defining a fixed and known current density (i.e., the ratio of current to the area from which current flows). Thus, if it is decided, based upon application of the above principles, that a specific maximum current for a given anode surface area e.g., I_{max}, will provide the enhanced efficiency drug delivery discussed above, then by increasing or decreasing the duty cycle, the amount of drug delivered at the high efficiency state can be increased or decreased without causing the applied current density to change. In choosing the parameters of drug delivery if using this approach, the magnitude of the current pulses is selected so that the resulting current density transforms the skin into the high efficiency state and the duty cycle of the current pulses is altered to adjust the drug delivery rate (i.e., a low dose of drug is administered by a high density (i.e., greater than I_c) pulsing current having a shorter pulse width, and hence a low duty cycle and a high dose of drug is administered by the same magnitude current density but being pulsed at a longer pulse width corresponding to a higher duty cycle.

This aspect of the invention is more specifically illustrated in Fig. 2 where waveforms for three different pulsing electrotransport currents of the same frequency are shown. In FIG. 2 time is illustrated on the horizontal axis, while current amplitude is illustrated on the vertical axis. The three current waveforms shown in FIG. 2 all have the same magnitude, and hence the same maximum applied current density I_{max} for an electrotransport delivery device of any one size. This particular current density I_{max} is greater than the critical current density level I_c. The three current waveforms have differing duty cycles, which is the percentage of time during which the current is applied. The three waveforms have duty cycles of 75% (top waveform), 50% (middle waveform) and 25% (bottom waveform). Thus, the 25% duty cycle waveform delivers drug transdermally by electrotransport at about

one-half the dosing level of the 50% duty cycle waveform and about one-third the dosing level of the 75% duty cycle waveform. All three waveforms administer drug transdermally by electrotransport through skin which is transformed into the high efficiency transfer state by reason of I_{max} being greater than I_c.

The second method of reducing the drug flux without reducing the applied level of electrotransport current, and hence current density, is to deliver the drug using a pulsing electrotransport current, the pulses of current producing a current density above I_c, and maintaining the pulse amptitude and pulse width constant while adjusting the pulsing frequency in order to control the drug delivery rate. In this manner, current density is maintained at or above the level which transforms the skin into the high efficiency state. Exemplary of this, a device employing a pulsed current waveform having current pulses with a magnitude of 0.2 mA, a pulse width of 10 msec, and a frequency of 10 Hz will deliver roughly half as much drug as the same device run at a frequency of 20 Hz. Given a constant drug delivery area, e.g., of an electrode assembly, the applied current densities of these two devices is the same and is above the high efficiency critical level I_c so that both devices deliver drug transdermally by electrotransport with higher efficiency and lower variability compared to devices which apply electrotransport current at current densities below the critical level I_c. From these two examples of the invention, one skilled in this art will appreciate that a combination of frequency and duty cycle may be used to alter the rate of drug delivery while maintaining the first average magnitude sufficient high to produce a current density above I_c. FIG. 4 shows the waveforms for a device operated to have a constant 9 msec pulse width, the frequency for a device operated according to the lower waveform being one-half that of a device operated according to the upper waveform (i.e., 50 Hz versus 100 Hz).

The third method of reducing the drug flux without reducing the applied level of electrotransport current, and hence current density, is to intentionally

deliver competitive co-ions (i.e., ionic species having a charge like that of the therapeutic agent, but which species do not induce a therapeutic effect when delivered into a patient) together with the desired drug so that some portion of the applied electrotransport current is carried by the co-ions rather than the drug ions. Delivery of competitive co-ions, for a given current, in addition to the drug or agent ions, provides adequate current density but reduces the quantity of therapeutic agent delivered. Delivery of competitive co-ions from, e.g., the drug reservoir, also reduces potentially expensive and potent total drug or agent delivered. This approach, under the specific conditions described, permits drug dosage control as well as providing enhanced stability of electrotransport therapeutic agent delivery efficiency. This approach is generally discouraged in the patent literature because it otherwise tends to reduce drug delivery efficiency. This aspect of this invention is particularly applicable to electrotransport delivery of those drugs or therapeutic agents which are therapeutically effective when (i) delivered at low transdermal fluxes and/or (ii) when present in low concentrations in the blood. Generally speaking, this aspect of the present invention is particularly applicable to the electrotransport delivery of highly potent drugs or other therapeutic agents.

The competitive ionic species can be loaded into the donor reservoir (e.g., a biocompatible salt is added to the donor reservoir) before electrotransport agent delivery and/or can be generated in situ during the operation of the electrotransport device. Generation of competitive ionic species in situ may be accomplished using a secondary electrode and appropriate electrical control circuitry as described in Phipps et al US Patent 5,443,442 for example.

The amount of the competitive species intentionally added to the donor reservoir will be specific to the drug or agents to be delivered and the relative electrophoretic mobilities of the drug ions and the competing ionic species. Generally, the competitive species will be ionic and should have delivery

characteristics similar to those of the drug being delivered. The quantity of co-delivered species to be added is selected so that the total current density is raised above the critical current density, I_c, where the ionic species efficiency is normalized or stabilized so that variation of delivery efficiency is no longer experienced.

The teachings in Theeuwes et al US Patent 5,080,646 may be utilized in determining the proper amount of competitive co-ion species to be added to the donor reservoir of an electrotransport delivery device. The patent discusses the processes involved in the transport of species through a biological surface such as skin, mucosa, or tissue. The Theeuwes et al Patent provides a mathematical analysis which permits one skilled in this art, when unacceptable random variability of electrically-assisted drug flux is experienced, to select a suitable quantity and species of competitive co-ion to be delivered along with the drug or agent.

Another way to use an inexpensive ionic species to transform the skin into the higher efficiency transfer state is to utilize a reverse polarity system wherein the electric current is initially applied at a level sufficient to produce a current density at or above I_c but which current carries the opposite polarity used to deliver the drug. In this way, the skin can be transferred into the higher/more stable efficiency state with application of current with little or no associated delivery of drug. Once the skin is transformed, the polarity of the applied electrotransport is then returned to the normal polarity used for drug delivery. One example of such a system has an anodic donor reservoir containing a cationic drug (D^+) with an anionic counter ion (X^-) such as chloride. The applied electrotransport current polarity is initially set to drive the counter ion X^- from the donor reservoir for at least the critical time, t_c, required to transform the skin to the high efficiency/stability state. Once the skin is transformed, the polarity of the applied current is reversed to deliver the drug cation D^+ from the donor reservoir into the skin.

As is noted above, agent delivery efficiency is preferably increased by exposure of the site to a current density at or above I_c and for a time period equal to or greater than a critical time, t_c. Generally speaking, for a pulsing electrotransport device, the pulse width (i.e., the length of the first segment of the waveform) must equal or exceed t_c. Thus, t_c, in a practice of this invention using pulsed current electrotransport devices and for delivery of fentanyl, falls between about 0.5 msec and 30 msec. It is believed that the minimum pulse width to cause transformation to the higher efficiency state is about 10 msec for fentanyl.

Table 1 shows data which support the above observation. Table 1 shows drug delivery efficiency data for a device programmed to run at frequencies of 1 Hz, 10 Hz and 625 Hz. A 31% duty cycle was employed.

TABLE 1

Frequency (Hz)	Pulse Width (msec)	Rate of Fentanyl Delivery (μg/hr)	
		Without Bolus Treatment	After Bolus Treatment*
625	0.5	7	34
10	31	52**	52**
1	310	48**	48**

* "Bolus Treatment" means a direct current bolus delivery of fentanyl for a period of 30 minutes at a current density of $0.1 mA/cm^2$.

** The numbers in these two columns are the same because even at a pulse width as short as 31 msec, the skin site had already transformed to its highly efficient state.

Table 1 also indicates that fentanyl delivery is significantly lower at a high pulsing frequency of 625 Hz compared to the lower pulsing frequencies

of 1 and 10 Hz. This phenomenon is called capacitive loss, which loss becomes greater as pulsing frequency is increased at a given duty cycle. Table 1 also shows that until a critical pulse width is achieved, regardless of frequency, no transformation of the body site agent delivery efficiency occurs.

Pulsed current electrotransport devices are well known in the art. Such devices are described in numerous technical articles and the patent literature including Bagniefski et al "A Comparison of Pulsed and Continuous Current Iontophoresis", Journal of Controlled Release, 113-122, (1090); McNichols et al, US patent 5,047,007; Sibalis US Patent 5,135,478; R. Burnette et al "Influence of Constant Current Iontophoresis on the Impedance and Passive Na^+ Permeability of Excised Nude Mouse Skin", 77 J.Pharmaceutical Sciences 492 (1988); Pikal et al, "Study of the Mechanisms of Flux Enhancement Through Hairless Mouse Skin by Pulsed DC Iontophoresis," 8 Pharmaceutical Research 365 (1991).

In a preferred aspect of the present invention, the efficiency (E) of transdermal electrotransport drug delivery is, at least at lower applied electrotransport current densities, dependent on the applied electrotransport current density (I_d). This phenomenon is illustrated graphically in FIG. 1. Specifically, when electrotransport current densities above a critical current density level, I_c, are applied to the skin of living animals for sufficient periods of time longer than a critical period of time, t_c, on the order of several milliseconds, the drug delivery efficiency (E) increases to a plateau level and is no longer dependent upon the level of applied current density. It is important to note that the variable electrotransport delivery efficiency effect is a limited exception to the widely reported principle that transdermal electrotransport drug flux is linearly dependent upon the level of applied electrotransport current. Our discovery is that this principle is only true at current densities above a critical current density level I_c. Thus, we have discovered that, at applied current densities below the critical current density level I_c, the rate of electrotransport drug delivery per unit of applied

electrotransport current is not constant as has been previously assumed. Not only is the electrotransport drug delivery efficiency (E) lower at current densities below I_c, E also exhibits greater variability at current densities below I_c than at current densities above the critical level I_c. Thus, at applied current densities below I_c, the electrotransport delivery is less efficient in that more electrotransport current must be applied to deliver a predetermined amount of drug. A still further aspect of our discovery is that the interpatient variability in transdermal electrotransport efficiency is lower at applied current densities above the critical level I_c and higher at applied current density levels below the critical level I_c.

In general, the critical current density level I_c for human skin is in the range of about 40 to 100 $\mu A/cm^2$, although the critical level I_c will vary somewhat depending upon (i) the particular drug being delivered, (ii) the particular patient being treated, and (iii) the particular skin location of the patient wearing the electrotransport device. Typically, a current density at or above the critical level I_c need only be applied for several milliseconds to several seconds before the skin enters the high efficiency drug transfer state. However, applied current densities below the critical level I_c are unable to transform the skin into the high efficiency transfer state, even when these low level current densities are applied for extended periods of time (e.g., up to several hours application). This transformation of the skin to a higher efficiency delivery state occurs only in living animals and does not occur with excised skin taken from living or dead animals, i.e., the skin transformation has not been found to occur when in vitro flux studies were run.

Once the skin has been transformed into the high efficiency transfer state, it tends to remain in that state for an extended period of time (e.g., up to 24 hours) even if no further electrotransport current is thereafter applied to the skin or if only low level current densities (i.e., current densities less than the critical level I_c) are thereafter applied to the skin. This result is

illustrated in FIG. 5 and is discussed below. The "transformed" skin is in general only those skin sites which are in contact with the donor and counter electrodes/reservoirs of the electrotransport delivery device and through which skin sites the applied current has been passed. Thus, if a skin site on the upper arm of a patient has been transformed by application of electrotransport current densities above the critical level I_c, the skin on the lower (same) arm, the legs, torso or other arm of the patient does not become transformed. The skin transformation of this invention is a localized phenomenon which is limited to those portions of the skin to which the donor and counter electrodes/reservoirs are attached. Since the skin at the counter electrode site also is converted to the high efficiency delivery state, methods and devices for delivering agents from the "donor" and "counter" electrodes, or both (e.g., by alternating current polarity) are within the scope of this invention.

Our discovery is particularly critical in those transdermal electrotransport drug delivery regimens wherein the drug is delivered at two (or more) different dosing levels, one dosing level being administered at a current density below the critical level I_c and another dosing level being administered at a current density above the critical level. For example, many drugs are adapted to be administered at a low dose baseline rate for extended periods, the baseline rate being interrupted periodically by periods of higher dosing. Examples of drugs which are administered in this fashion include (1) analgesics, such as fentanyl and sufentanil, which are administered at a low baseline level to treat (e.g., chronic) pain and which are periodically delivered at higher doses to treat more severe episodes of pain; (2) anti-emetics, such as the 5HT3 receptor antagonists ondansetron and granisetron, which are administered continuously at low levels (e.g., during weeks over which a patient is undergoing chemotherapy) and which are periodically administered at higher dosing levels (i.e., during the actual chemotherapeutic administration); (3) anti-epileptics, such as phenytoin,

which are delivered continuously at low baseline levels and periodically at higher levels when the patient is undergoing an epileptic seizure; and (4) anti-diabetic drugs, such as insulins, which can be delivered continuously at low baseline levels and periodically (e.g., just before, during or after meals) at higher levels. The problem encountered with this type of transdermal electrotransport drug administration is that after the drug is administered at the higher dosing rate (with the applied current density above the critical level, I_c), when the applied electrotransport current is readjusted to apply the original lower baseline level, the transdermal electrotransport drug flux does not return to the same baseline level. The drug flux instead falls to a level somewhere between the original baseline rate and the high dosing rate, because the skin has been transformed into a higher efficiency drug delivery state. For example, if the efficiency is enhanced by a factor of two, after the skin has experienced a current density above the critical current density, and then the current is lowered to the original base line current, the drug delivery rate would be twice that experienced before the transformation. The higher baseline rate could result in a drug overdose if the electrotransport system does not compensate for this shift in efficiency. To eliminate this problem, the electrotransport system should reduce the current applied (e.g., by approximately a factor of two) after the skin has experienced a current density greater than I_c. With reference to FIG. 1, data point 2 is a likely efficiency that would be experienced at the drug delivery site were current (and therefore current density) reduced after exposure of the body site to current density at or above I_c for at least a period of time t_c. At data point "2" electrotransport agent delivery efficiency is higher than the agent delivery efficiency which was initially experienced at a current density of about 20 $\mu A/cm^2$ (i.e., at a time before exposure of the skin to a current density above I_c).

Reference is now made to FIG. 3 which depicts an exemplary electrotransport device which can be used in accordance with the present

invention. FIG. 3 shows a perspective exploded view of an electrotransport device 10 having an activation switch in the form of a push button switch 12 and a display in the form of a light emitting diode (LED) 14. Device 10 comprises an upper housing 16, a circuit board assembly 18, a lower housing 20, anode electrode 22, cathode electrode 24, anode reservoir 26, cathode reservoir 28 and skin-compatible adhesive 30. Upper housing 16 has lateral wings 15 which assist in holding device 10 on a patient's skin. Upper housing 16 is preferably composed of an injection moldable elastomer (e.g., ethylene vinyl acetate). Printed circuit board assembly 18 comprises an integrated circuit 19 coupled to discrete electrical components 40 and battery 32. Circuit board assembly 18 is attached to housing 16 by posts (not shown in FIG. 3) passing through openings 13a and 13b, the ends of the posts being heated/melted in order to heat stake the circuit board assembly 18 to the housing 16. Lower housing 20 is attached to the upper housing 16 by means of adhesive 30, the upper surface 34 of adhesive 30 being adhered to both lower housing 20 and upper housing 16 including the bottom surfaces of wings 15.

Shown (partially) on the underside of circuit board assembly 18 is a battery 32, which is preferably a button cell battery and most preferably a lithium cell. Other types of batteries, such as sizes AAA and AAAA may also be employed to power device 10.

The circuit outputs (not shown in FIG. 3) of the circuit board assembly 18 make electrical contact with the electrodes 24 and 22 through openings 23,23' in the depressions 25,25' formed in lower housing, by means of electrically conductive adhesive strips 42,42'. Electrodes 22 and 24, in turn, are in direct mechanical and electrical contact with the top sides 44',44 of drug reservoirs 26 and 28. The bottom sides 46',46 of drug reservoirs 26,28 contact the patient's skin through the openings 29',29 in adhesive 30. Upon depression of push button switch 12, the electronic circuitry on circuit board assembly 18 delivers a predetermined DC current to the electrodes/reservoirs

22,26 and 24,28 for a delivery interval of predetermined length, e.g., about 10 minutes. Preferably, the device transmits to the user a visual and/or audible confirmation of the onset of the drug delivery, or bolus, interval by means of LED 14 becoming lit and/or an audible sound signal from, e.g., a "beeper". Drug (e.g., an analgesic drug such as fentanyl) is then delivered through the patient's skin, e.g., on the arm, for the predetermined delivery interval. In practice, a user receives feedback as to the onset of the drug delivery interval by visual (LED 14 becomes lit) and/or audible signals (a beep from the "beeper"). A preferred device is described in commonly owned, pending patent application entitled "Display for an Electrotransport Device", US Patent Application Serial Number 08/410,112, filed March 24, 1995.

Anodic electrode 22 is preferably comprised of silver and cathodic electrode 24 is preferably comprised of silver chloride. Both reservoirs 26 and 28 are preferably comprised of polymer hydrogel materials as described herein. Electrodes 22, 24 and reservoirs 26, 28 are retained by lower housing 20. When the drug being delivered by electrotransport is cationic, the anodic reservoir 26 is the "donor" reservoir which contains the drug and the cathodic reservoir 28 contains a biocompatible electrolyte. When the drug being delivered by electrotransport is anionic, the cathodic reservoir 28 is the "donor" reservoir which contains the drug and the anodic reservoir 26 contains a biocompatible electrolyte.

The push button switch 12, the electronic circuitry on circuit board assembly 18 and the battery 32 are adhesively "sealed" between upper housing 16 and lower housing 20. Upper housing 16 is preferably composed of rubber or other elastomeric material. Lower housing 20 is preferably composed of a plastic or elastomeric sheet material (e.g., polyethylene) which can be easily molded to form depressions 25,25' and cut to form openings 23,23'. The assembled device 10 is preferably water resistant (i.e., splash proof) and is most preferably waterproof. The system has a low profile that easily conforms to the body thereby allowing freedom of

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The push button switch 12, the electronic circuitry on circuit board assembly 18 and the battery 32 are adhesively "sealed" between upper housing 16 and lower housing 20. Upper housing 16 is preferably composed of rubber or other elastomeric material. Lower housing 20 is preferably composed of a plastic or elastomeric sheet material (e.g., polyethylene) which can be easily molded to form depressions 25,25' and cut to form openings 23,23'. The assembled device 10 is preferably water resistant (i.e., splash proof) and is most preferably waterproof. The system has a low profile that easily conforms to the body thereby allowing freedom of

movement at, and around, the wearing site. The anode reservoir 26 and the cathode reservoir 28 are located on the skin-contacting side of device 10 and are sufficiently separated to prevent accidental electrical shorting during normal handling and use.

The device 10 adheres to the patient's body surface (e.g., skin) by means of a peripheral adhesive 30 which has upper side 34 and body-contacting side 36. The adhesive side 36 has adhesive properties which assures that the device 10 remains in place on the body during normal user activity, and yet permits reasonable removal after the predetermined (e.g., 24-hour) wear period. Upper adhesive side 34 adheres to lower housing 20 and retains the electrodes and drug reservoirs within housing depressions 25,25' as well as retains lower housing 20 attached to upper housing 16.

The push button switch 12 is located on the top side of device 10 and is easily actuated through clothing. A double press of the push button switch 12 within a short period of time, e.g., three seconds, is preferably used to activate the device 10 for delivery of drug, thereby minimizing the likelihood of inadvertent actuation of the device 10.

Upon switch activation an audible alarm signals the start of drug delivery, at which time the circuit supplies a predetermined level of DC current to the electrodes/reservoirs for a predetermined delivery interval. The LED 14 remains "on" throughout the delivery interval indicating that the device 10 is in an active drug delivery mode. The battery preferably has sufficient capacity to continuously power the device 10 at the predetermined level of DC current for the entire wearing period.

The present invention is particularly useful in the transformation of human skin in the transdermal electrotransport delivery of drugs to humans. However, the invention also has utility in delivering drugs to other animals and is not limited to humans.

The terms "agent" and "drug" are used interchangeably herein and are intended to have their broadest interpretation as any therapeutically active substance which is delivered to a living organism to produce a desired, usually beneficial, effect. In general, this includes therapeutic agents in all of the major therapeutic areas including, but not limited to, anti-infectives such as antibiotics and antiviral agents, analgesics and analgesic combinations, anesthetics, anorexics, antiarthritics, antiasthmatic agents, anticonvulsants, anti-depressants, antidiabetic agents, antidiarrheals, antihistamines, anti-inflammatory agents, antimigraine preparations, antimotion sickness preparations, antinauseants, antineoplastics, antiparkinsonism drugs, antipruritics, antipsychotics, antipyretics, antispasmodics including gastrointestinal and urinary antispasmodics, anticholinergics, sympathomimetrics, xanthine derivatives, cardiovascular preparations including calcium channel blockers, beta-blockers, antiarrythmics, antihypertensives, diuretics, vasodilators including general, coronary, peripheral and cerebral vasodilators, central nervous system stimulants, cough and cold preparations, decongestants, diagnostics, hormones, hypnotics, immunosuppressives, muscle relaxants, parasympatholytics, parasympathomimetrics, proteins, peptides, polypeptides and other macromolecules, psychostimulants, sedatives and tranquilizers.

The present invention can be used to deliver transdermally by electrotransport the following drugs: interferons, alfentanyl, amphotericin B, angiopeptin, baclofen, beclomethasone, betamethasone, bisphosphonates, bromocriptine, buserelin, buspirone, calcitonin, ciclopirox, olamine, copper, cromolyn sodium, desmopressin, diclofenac diflorasone, diltiazem, dobutamine, dopamine agonists, dopamine agonists, doxazosin, droperidol, enalapril, enalaprilat, fentanyl, encainide, G-CSF, GM-CSF, M-CSF, GHRF, GHRH, gonadorelin, goserelin, granisetron, haloperidol, hydrocortisone, indomethacin, insulin, insulinotropin, interleukins, isosorbide dinitrate, ketoprofen, ketorolac, leuprolide, LHRH, lidocaine, lisinopril, LMW heparin,

melatonin, methotrexate, metoclopramide, miconazole, midazolam, nafarelin, nicardipine, NMDA antagonists, octreotide, ondansetron, oxybutynin, PGE_1, piroxicam, pramipexole, prazosin, prednisolone, prostaglandins, scopolamine, seglitide, sufentanil, terbutaline, testosterone, tetracaine, tropisetron, vapreotide, vasopressin, verapamil, warfarin, zacopride, zinc, and zotasetron.

This invention is also believed to be useful in the transdermal electrotransport delivery of peptides, polypeptides and other macromolecules typically having a molecular weight of at least about 300 daltons, and typically a molecular weight in the range of about 300 to 40,000 daltons. Specific examples of peptides and proteins in this size range include, without limitation, LHRH, LHRH analogs such as buserelin, gonadorelin, nafarelin and leuprolide, GHRH, insulin, heparin, calcitonin, endorphin, TRH, NT-36 (chemical name: N=[[(s)-4-oxo-2-azetidinyl]carbonyl]-L-histidyl-L-prolinamide), liprecin, pituitary hormones (e.g., HGH, HMG, HCG, desmopressin acetate, etc.,), follicle luteoids, αANF, growth hormone releasing factor (GHRF), βMSH, TGF-β, somatostatin, atrial natriuretic peptide, bradykinin, somatotropin, platelet-derived growth factor, asparaginase, bleomycin sulfate, chymopapain, cholecystokinin, chorionic gonadotropin, corticotropin (ACTH), epidermal growth factor, erythropoietin, epoprostenol (platelet aggregation inhibitor), follicle stimulating hormone, glucagon, hirulogs, hyaluronidase, interferons, insulin-like growth factors, interleukins, menotropins (urofollitropin (FSH) and LH), oxytocin, streptokinase, tissue plasminogen activator, urokinase, vasopressin, ACTH analogs, ANP, ANP clearance inhibitors, angiotensin II antagonists, antidiuretic hormone agonists, antidiuretic hormone antagonists, bradykinin antagonists, CD4, ceredase, CSF's, enkephalins, FAB fragments, IgE peptide suppressors, IGF-1, neuropeptide Y, neurotrophic factors, opiate peptides, parathyroid hormone and agonists, parathyroid hormone antagonists, prostaglandin antagonists, pentigetide, protein C, protein S, ramoplanin, renin

inhibitors, thymosin alpha-1, thrombolytics, TNF, vaccines, vasopressin antagonist analogs, alpha-1 anti-trypsin (recombinant).

Generally speaking, it is most preferable to use a water soluble form of the drug or agent to be delivered. Drug or agent precursors, i.e., species which generate the selected species by physical or chemical processes such as ionization, dissociation, dissolution or covalent chemical modification (i.e., prodrugs), are within the definition of "agent" or "drug" herein. "Drug" or "agent" is to be understood to include charged and uncharged species as described above.

While the disclosure has focused upon the electrotransport delivery of ionic species, the present invention is also applicable to the electrotransport delivery of uncharged species, e.g., by electroosmosis. Thus, the transformation of the skin into the high efficiency transport state is not limited to electrically assisted transport of ionic species but also to electroosmotic delivery of uncharged (i.e., non-ionized) species.

The following examples illustrate some of the advantages of the present invention.

EXAMPLE 1

Pulsing Frequency and Fentanyl Flux

This study assessed the effect of pulsing frequency on the electrotransport delivery of fentanyl using pulsed current waveforms. The frequencies evaluated in this study were 1, 10, and 625 Hz.

The electrotransport devices were configured to deliver a 200 μA square wave current pulse, having a 31% duty cycle. At the frequency of 1 Hz, the 31% duty cycle square wave current achieved a current pulse width of 310 msec. At the frequency of 10 Hz, the 31% duty cycle square wave current achieved a current pulse width of 31 msec. At the frequency of 625 Hz, the 31% duty cycle square wave current achieved a current pulse

width of 0.5 msec. The electrotransport devices delivered fentanyl through the skin from a donor hydrogel having a skin contact surface area of 2 cm^2. Thus, the applied maximum current density, I_{max}, was 100 $\mu A/cm^2$ (200 $\mu A \div 2\ cm^2 = 100\ \mu A/cm^2$). The gels were imbibed with an aqueous solution of fentanyl HCl. After treatment periods of varying duration, the electrotransport devices were removed. The skin site was then washed to remove any residual fentanyl.

For each treatment, blood samples were taken commencing with the application of current from the electrotransport devices. Serum fentanyl levels from each patient were used to calculate mean drug flux.

FIG. 7 shows that the use of a square-wave frequency of 625 Hz resulted in minimal fentanyl flux. This is shown in the lower most nearly horizontal curve in FIG. 7. The use of the lower pulsing frequencies, 1 and 10 Hz, resulted in increased fentanyl flux. This is shown in the upper two curves of FIG. 7. No statistically significant difference in fentanyl flux was observed between 1 and 10 Hz. These results suggest that the use of lower pulsing frequencies results in higher electrotransport delivery efficiency of fentanyl.

The remaining Examples do not utilize a pulsing electrotransport current, and are therefore relevant only to the preferred aspect of the present invention wherein the applied current density (of each of the pulses) is maintained above I_c.

EXAMPLE 2

Current Density and Increased Efficiency

This study evaluated the effect of applied current density on electrotransport drug delivery efficiency. Drug delivery efficiency is expressed in terms of the rate of drug delivery per unit of applied current. The study

involved the application of electrotransport devices to eighteen healthy male volunteers for a duration of about one day.

The two electrotransport treatments involved the delivery of fentanyl, from a donor reservoir containing an aqueous solution of fentanyl HCl and having a skin-contact area of 5 cm^2, at a baseline current of 100 μA. Thus, the applied electrotransport current density was 20 μA/cm^2 (= 100 μA ÷ 5 cm^2). Six of the eighteen volunteers were administered 4 bolus doses during the first hour of treatment by applying current levels of 1300 μA (i.e., an applied electrotransport current density of 260 μA/cm^2) for a duration of 2.5 minutes at 15 minute intervals. Following the administration of the four boluses in the first hour of treatment, these six volunteers received continuous transdermal electrotransport fentanyl administration at a current density of 20 μA/cm^2 from hour 2 through 24 hours. The remaining twelve volunteers received continuous transdermal electrotransport fentanyl administration at a current density of 20 μA/cm^2 over the entire 24 hour delivery period. After the treatment period, the electrotransport devices were removed. The skin site was then washed to remove any residual fentanyl.

Blood samples were taken over the entire 24 hour period commencing with the application of current from the electrotransport devices. Serum fentanyl concentrations were used to calculate mean transdermal fentanyl fluxes using subject specific pharmacokinetic parameters and conventional methods.

FIG. 5 shows that once a skin site receives a minimum level of current (for a fixed electrode area) for a sufficient duration, a high electrotransport efficiency state is achieved. FIG. 5 shows the mean serum fentanyl concentration in the blood of the subjects over the 24 hour testing period. As is shown in the uppermost curve (◊•••◊•••◊) in FIG. 5, the six volunteers which received the four 260 μA/cm^2, 2.5 minute bolus administrations in the first hour of treatment exhibited higher efficiency fentanyl transdermal delivery than the group of twelve subjects shown as three groups of four in the three

lower curves (to emphasize inherent variability) who received only the 20 μA/cm^2 constant DC current. Once this high-efficiency transport state is achieved, more drug is delivered through the skin per unit of applied current. Further, the effect lasted the entire 24 hours of the treatment. This is indicated by the vertical separation between the upper curve and the three lower curves in FIG. 5.

Specifically, the six volunteers who received the four 260 μA/cm^2 doses in the first hour of treatment exhibited a mean transdermal fentanyl flux of 113 μg/h while the twelve volunteers who received only the 20 μA/cm^2 baseline current exhibited a mean transdermal fentanyl flux of 57 μg/h. This indicates that the efficiency was enhanced by about a factor of two as a result of the initial high current density applications.

EXAMPLE 3

Current Density and Fentanyl Flux

This study was undertaken to evaluate the relationship of current density and drug flux in the transdermal electrotransport delivery of fentanyl. Electrotransport devices, delivering constant DC currents, were applied to 8 healthy male volunteers for a duration of 24 hours. The three electrotransport treatment regimens in this study differed only in the applied electrotransport current (and therefore current density) levels. The electrotransport devices delivered fentanyl through the skin from a donor hydrogel having a skin contact surface area of 5 cm^2. The gels were imbibed with an aqueous solution of fentanyl HCl. The current density levels used in this study were 10, 20, and 40 μA/cm^2. After a 24 hour treatment period, the electrotransport devices were removed. The skin site was then washed to remove any residual fentanyl. All 8 volunteers received each treatment approximately 1 week apart.

For each treatment, blood samples were taken over a 24 hour period commencing with the application of current from the electrotransport devices. Serum fentanyl concentrations over the 24 hours are shown in FIG. 6. The top curve (--Δ--Δ--Δ--) in FIG. 6 was the 200 μA treatment (i.e., 40 μA/cm^2), the middle curve (-□-□-□-) the 100 μA treatment (i.e., 20 μA/cm^2) and the bottom curve (--O--O--O--) the 50 μA treatment (i.e., 10 μA/cm^2). As in Example 2, the serum fentanyl concentrations from each patient were used to calculate mean transdermal fentanyl flux and the mean total amount of fentanyl delivered. A drug delivery efficiency level for each treatment was derived by dividing the mean fentanyl delivery rate by the current density applied to the skin.

The average transdermal fentanyl fluxes were 19, 73 and 173 μg/h at the applied current densities 10, 20 and 40 μA/cm^2, respectively. This data shows a non-linear relationship between applied current and transdermal electrotransport fentanyl flux within the electrotransport current density range of 10 to 40 μA/cm^2. An almost ten-fold increase in drug delivery rate was observed as the current was increased four-fold from 50 μA to 200 μA. This unexpected result indicates that the efficiency of fentanyl delivery was enhanced by a factor of about 2.5-fold due to the change in current density from 10 to 40 μA/cm^2.

EXAMPLE 4

This study was undertaken to evaluate the relationship between current density and drug flux in the transdermal electrotransport delivery of goserelin. The study involved the application of electrotransport devices, applying constant current, to 12 normal male volunteers for a duration of 8 hours.

The two electrotransport treatment regimens in this study differed only in applied current density levels. The electrotransport devices delivered goserelin through the skin from polyvinyl alcohol (PVOH)-based donor hydrogels having a skin-contact surface area of 4 cm^2. The gels contained an aqueous goserelin solution. The current density levels used in this study were 50 and 100 $\mu A/cm^2$. After an 8 hour treatment period, the electrotransport devices were removed. The skin site was then washed to remove any residual goserelin. All 12 volunteers received each treatment seven days apart.

For each treatment, seven blood samples were taken over a 24 hour period commencing with the application of current from the electrotransport devices. Serum goserelin concentrations from each patient were used to calculate mean drug flux and the mean total amount of drug delivered.

FIG. 8 shows the goserelin blood plasma concentrations for the 8 hour duration of electrotransport administration for the two current densities (i.e., 50 and 100 $\mu A/cm^2$). The 100 $\mu A/cm^2$ curve is the upper curve in FIG. 8 while the lower curve in FIG. 8 is the 50 $\mu A/cm^2$ data. From this concentration data, transdermal goserelin fluxes were calculated. The average transdermal goserelin flux was 5.8 μg/h at an applied current density of 50 $\mu A/cm^2$ while the average transdermal flux of goserelin was 21.6 μg/h at an applied current density of 100 $\mu A/cm^2$. Thus, a non-linear relationship between applied current density and drug flux was shown by the data. An almost four-fold increase in drug flux is observed as the current density rises from 50 to 100 $\mu A/cm^2$. This data also suggests the existence of a critical current density, I_c, which for transdermal electrotransport delivery of goserelin falls between 50 and 100 $\mu A/cm^2$, above which more drug is delivered through the skin per unit of applied current.

The above disclosure will suggest many alternatives, permutations, and variations of the invention to one skilled in this art without departing from the scope of the invention. The above disclosure is intended to be illustrative

and not exhaustive. All such, permutations, variations, and alternatives suggested by the above disclosure are to be included within the scope of the attached claims.

Claims:

1. An device (10) for delivering a therapeutic agent through a body surface by electrotransport, the device (10) having a donor reservoir (26,46) containing the therapeutic agent, the donor reservoir (26,46) being adapted to be placed in therapeutic agent-transmitting relation with the body surface, the device (10) also having a source of electrical power (32) and a current controller (19,40), the power source (32) and current controller (19,40) being adapted to apply a pulsing electrotransport current to the reservoir (26,46) and the body surface, the pulsing electrotransport current having a periodic waveform with a first segment of applied electrotransport current having a first average current magnitude, and a second segment of applied electrotransport current having a second average current magnitude, the second average current magnitude being less than the first average current magnitude, the device (10) being characterized by:

the first segment having a length of at least 5 msec.

2. The device of claim 1, wherein the length of the first segment is at least 10 msec.

3. The device of claim 1, wherein the electrotransport current has a pulsing frequency of less than about 100 Hz.

4. The device of claim 1, wherein the electrotransport current has a pulsing frequency of less than about 10 Hz.

5. The device of claim 1, wherein the first segment has a maximum current magnitude, which provides a maximum current density I_{max}.

6. The device of claim 5, wherein I_{max} is greater than or equal to 40 μA/cm^2.

7. The device of claim 1, wherein the first average current magnitude provides an average current density greater than or equal to 40 μA/cm^2.

8. The device of claim 1, wherein the device (10) is adapted to be applied to skin of a human patient.

9. The device of claim 1, wherein the therapeutic agent is fentanyl, the controller (19,40) controls the first average current magnitude to provide an average current density of at least 40 μA/cm^2 during the first segment, and the controller (19,40) controls the length of the first segment to at least about 10 msec.

10. The device of claim 1, wherein the therapeutic agent is goserelin, the controller (19,40) controls the first average current magnitude to provide an average current density of at least about 50 μA/cm^2 during the first segment, and the controller (19,40) controls the length of the first segment to at least about 10 msec.

11. The device of claim 1, wherein the controller (19,40) can adjust the relative lengths of the first and second segments in order to vary the therapeutic agent delivery rate.

12. The device of claim 1, wherein the donor reservoir (26,46) contains an intentionally added competitive co-ion species whereby the device (10) delivers the agent through the body surface at a reduced rate.

13. The device of claim 1, wherein the second average current magnitude is substantially zero.

14. The device of claim 13, wherein the pulsing current has a square waveform shape.

15. A method of operating an electrotransport delivery device (10) delivering a therapeutic agent through a body surface by electrotransport, including controlling electrotransport current applied by the device (10) to be a pulsing electrotransport current, the pulsing electrotransport current having a periodic waveform with a first segment of applied electrotransport current having a first average current magnitude, and a second segment of applied electrotransport current having a second average current magnitude, the second average current magnitude being less than the first average current magnitude, the method characterized by:

controlling the length of the first segment to at least 5 msec.

16. The method of claim 15, including controlling the length of the first segment to at least 10 msec.

17. The method of claim 15, wherein the pulsing current has a pulsing frequency of less than about 100 Hz.

18. The method of claim 15, wherein the pulsing current has a pulsing frequency of less than about 10 Hz.

19. The method of claim 15, wherein the first segment has a maximum current magnitude, which provides a maximum current density I_{max}.

20. The method of claim 19, wherein I_{max} is greater than or equal to 40 $\mu A/cm^2$.

21. The method of claim 16, wherein the first average current magnitude provides a current density greater than or equal to 40 $\mu A/cm^2$.

22. The method of claim 16, wherein the agent is fentanyl, the body surface is human skin, the first average current magnitude is controlled to provide an average current density of at least about 40 $\mu A/cm^2$, and the segment of applied electric current is controlled to be at least about 10 msec.

23. The method of claim 16, wherein the agent is goserelin, the body surface is human skin, the first average current magnitude is controlled to provide an average current density of at least about 50 $\mu A/cm^2$, and the segment of applied electric current is controlled to be at least about 10 msec.

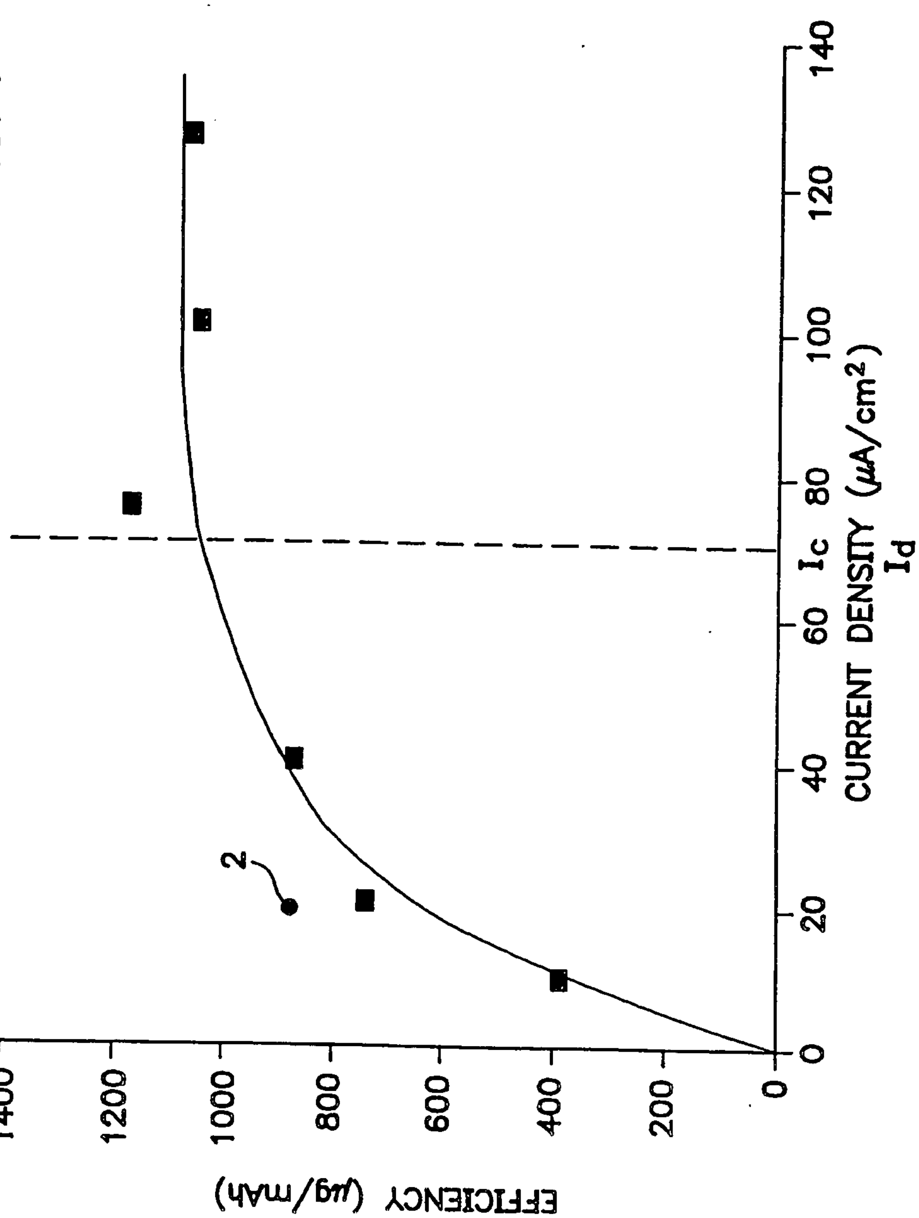
24. The method of claim 16, including the step of adjusting the relative lengths of the first and second segments to vary the therapeutic agent delivery rate.

25. The method of claim 16, including intentionally adding a competitive co-ion species to the donor reservoir (26,46), whereby the device (10) delivers the therapeutic agent through the body surface at a reduced rate.

26. The method of claim 16, wherein the second average current magnitude is substantially zero.

27. The method of claim 26, wherein the pulsing current has a square waveform shape.

FIG. I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1000
800
600
400
200
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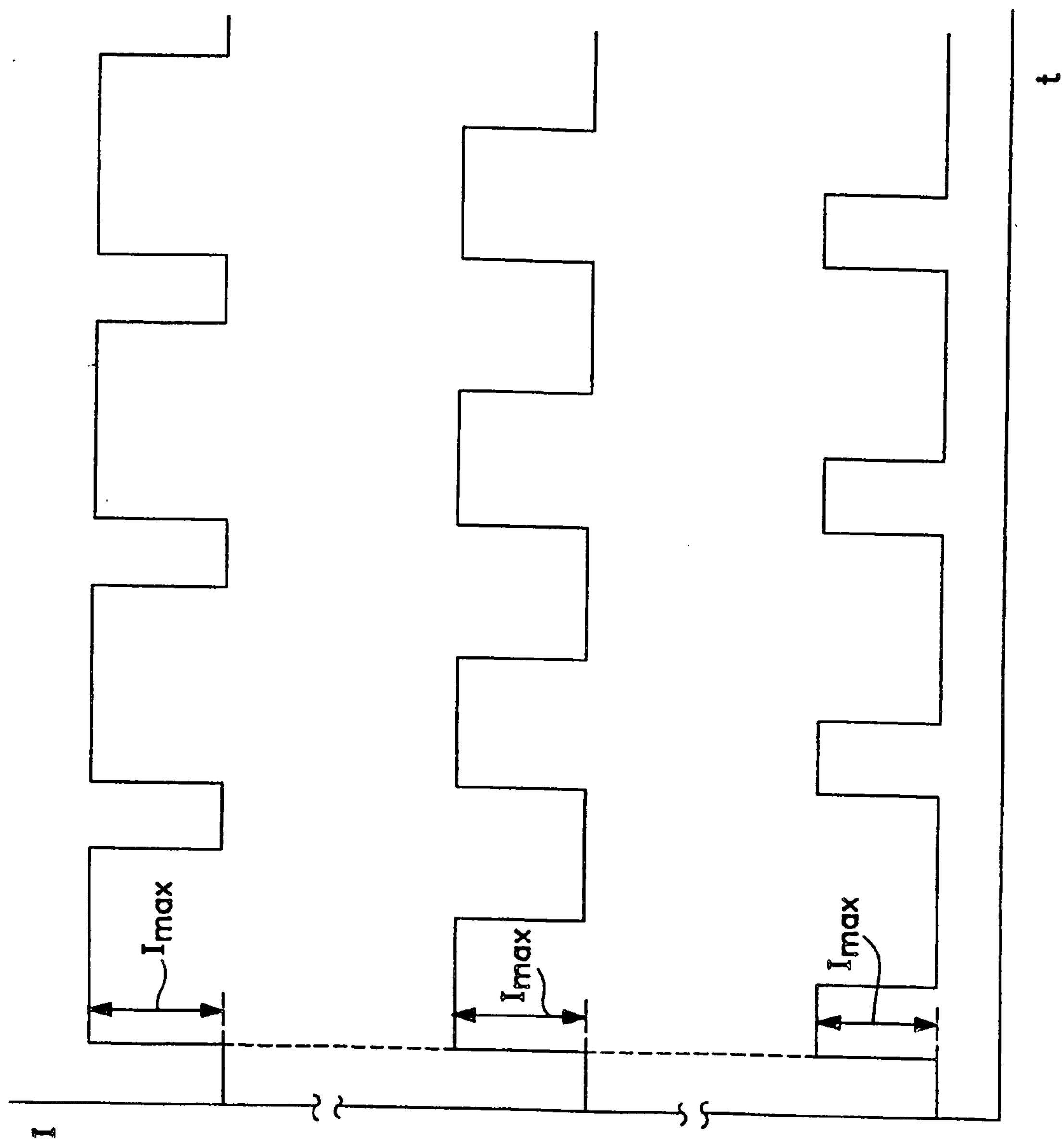
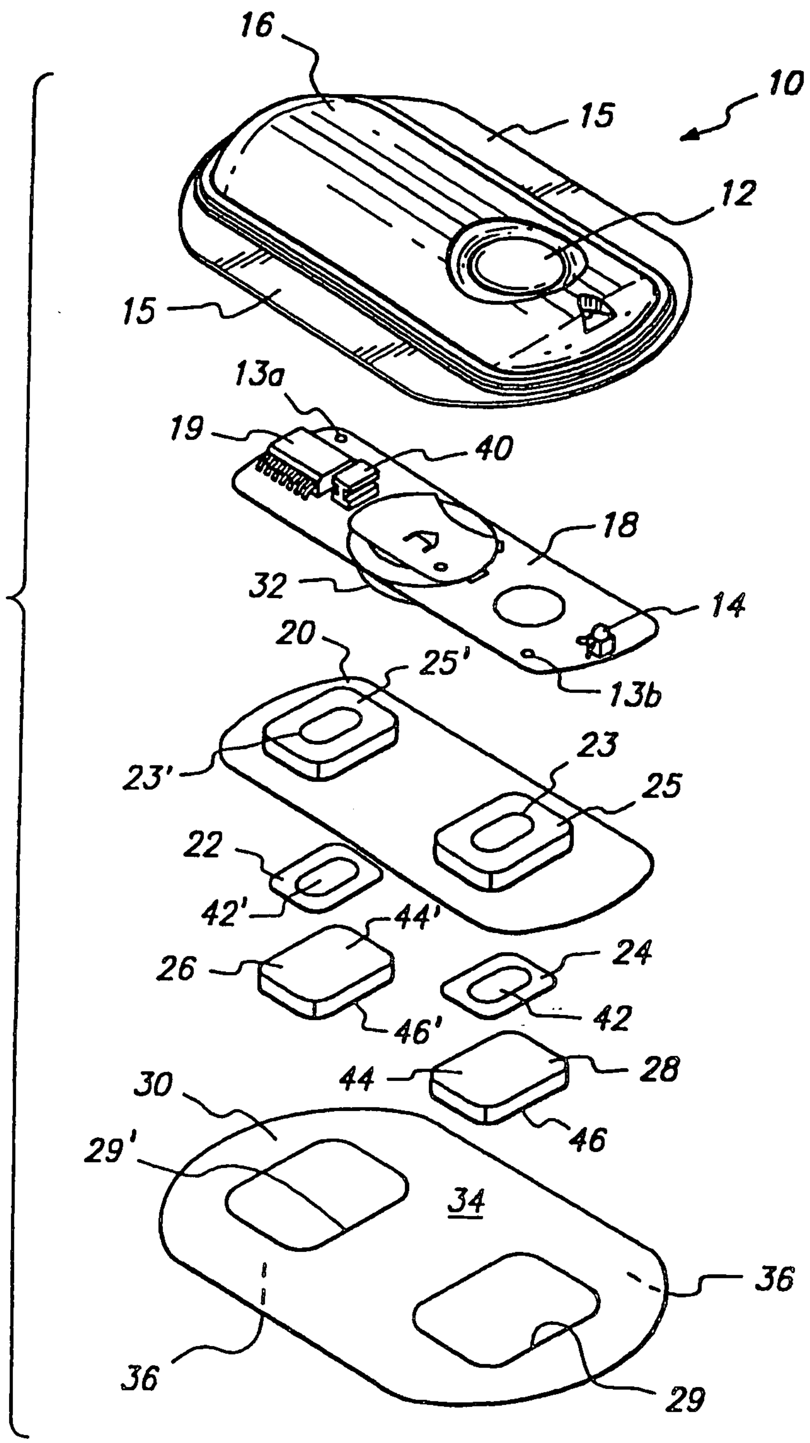
0
20
40
60
80
100
120
140
CURRENT DENSITY (µA/cm²)
Ic
Id
2

FIG. 2

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14
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25'
13b
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23'
25
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42'
44'
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26
42
46'
44
28
30
46
29'
34
36
36
29

FIG. 3

FIG. 4

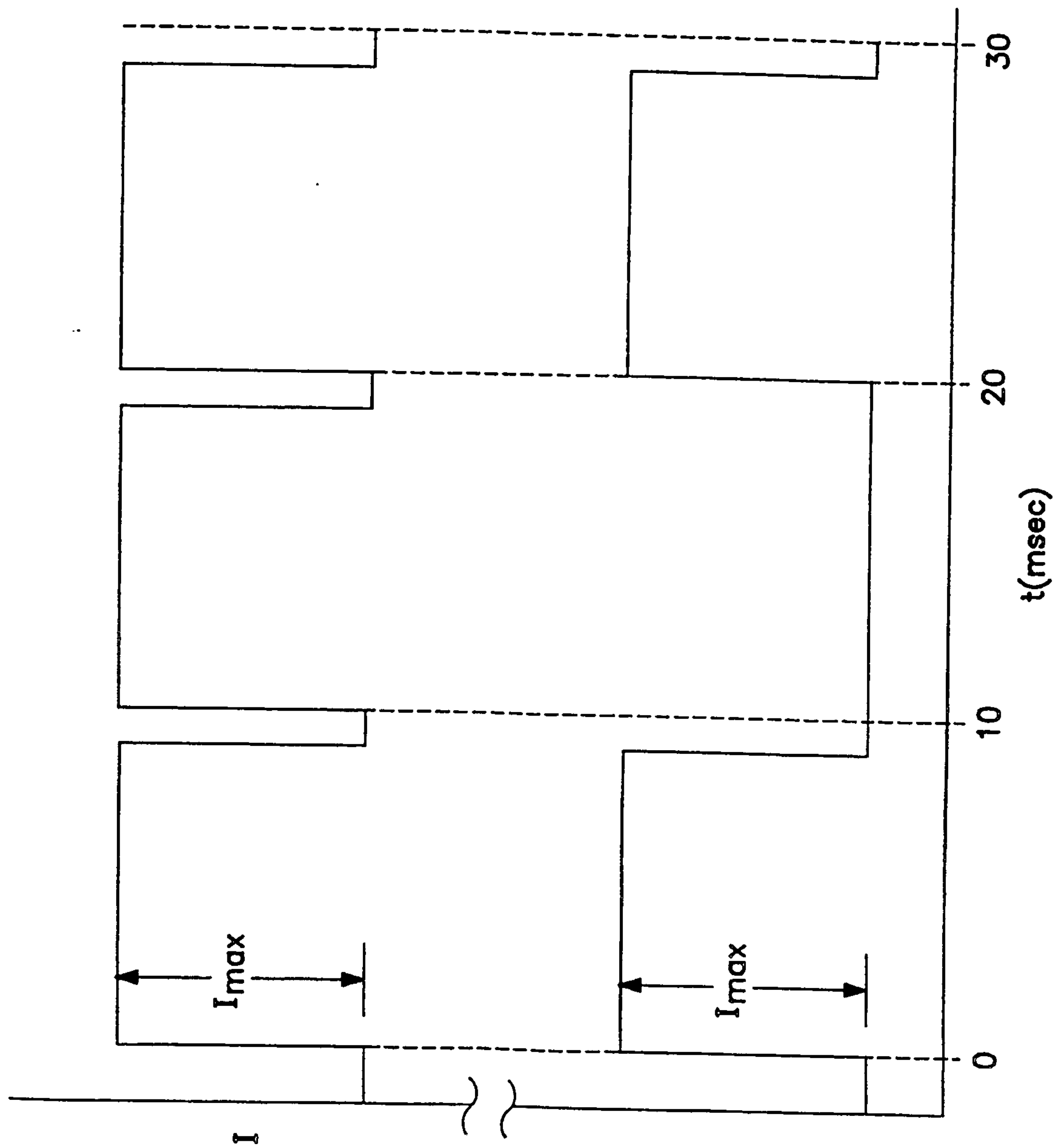
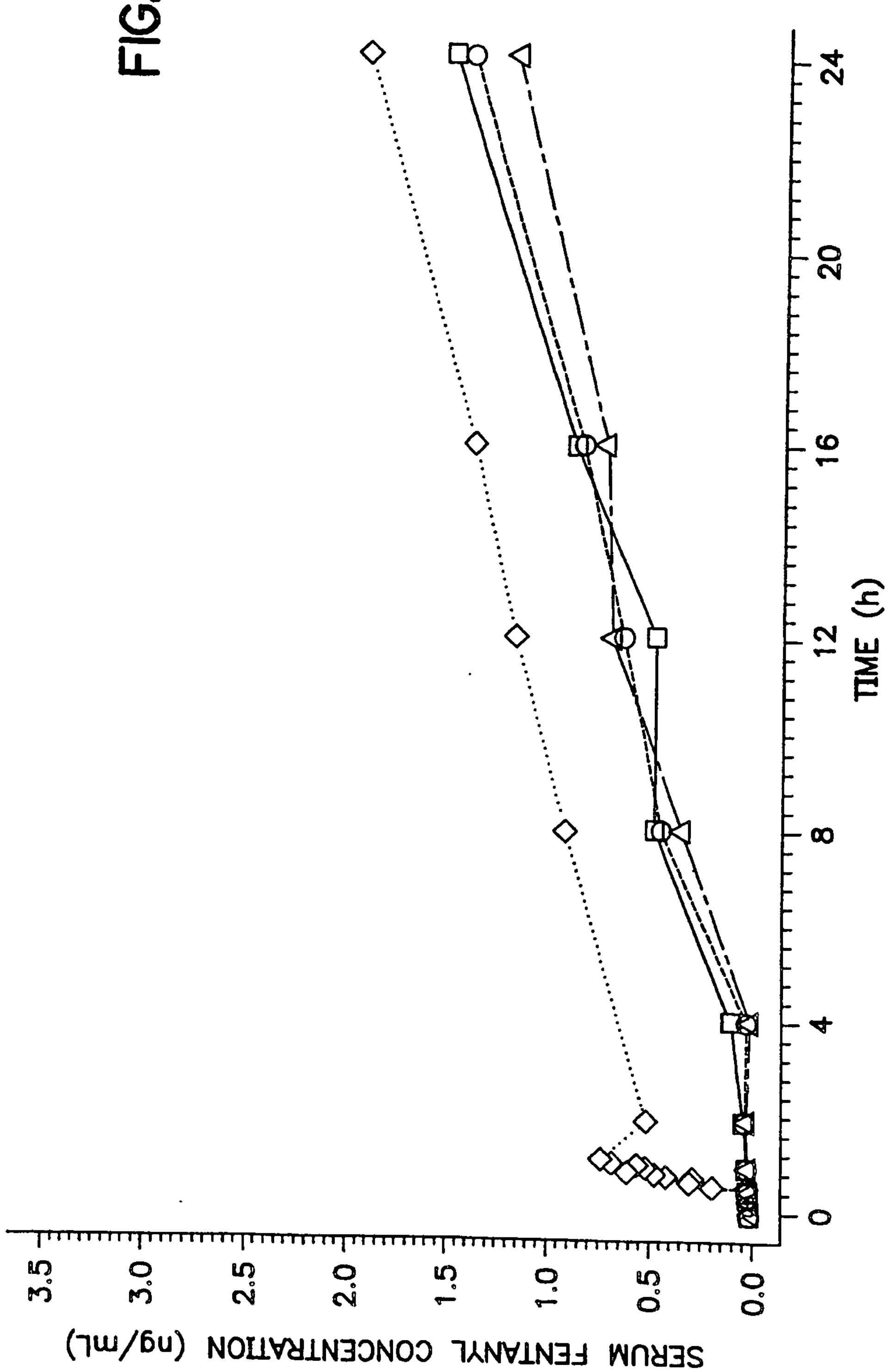


FIG. 5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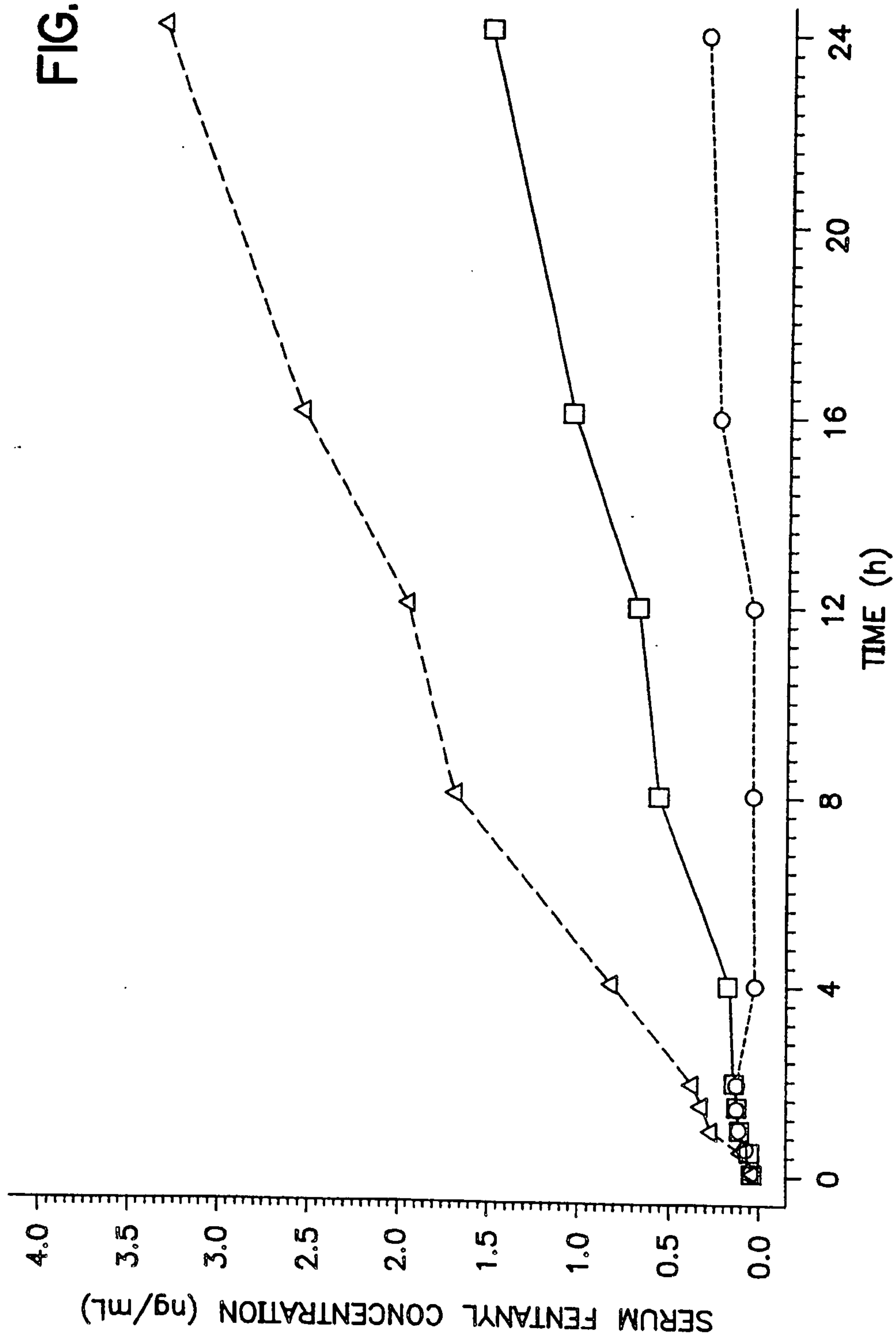
0.5
1.0
1.5
2.0
2.5
3.0
3.5
TIME (h)
0
4
8
12
16
20
24

FIG. 6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0.5
1.0
1.5
2.0
2.5
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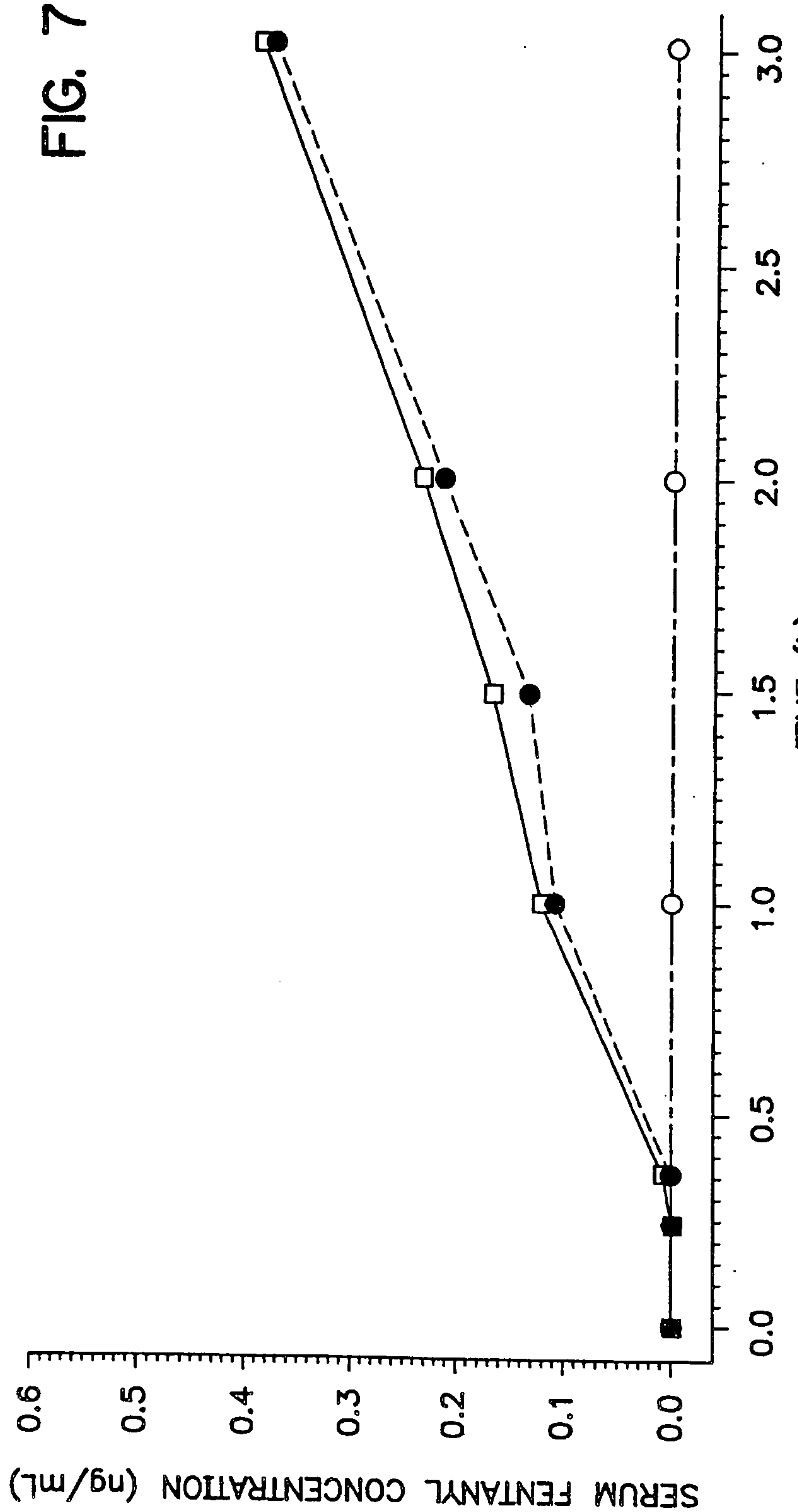
4.0
TIME (h)
0
4
8
12
16
20
24

FIG. 7


SERUM FENTANYL CONCENTRATION (ng/mL)
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0.4
0.3
0.2
0.1
0.0
TIME (h)
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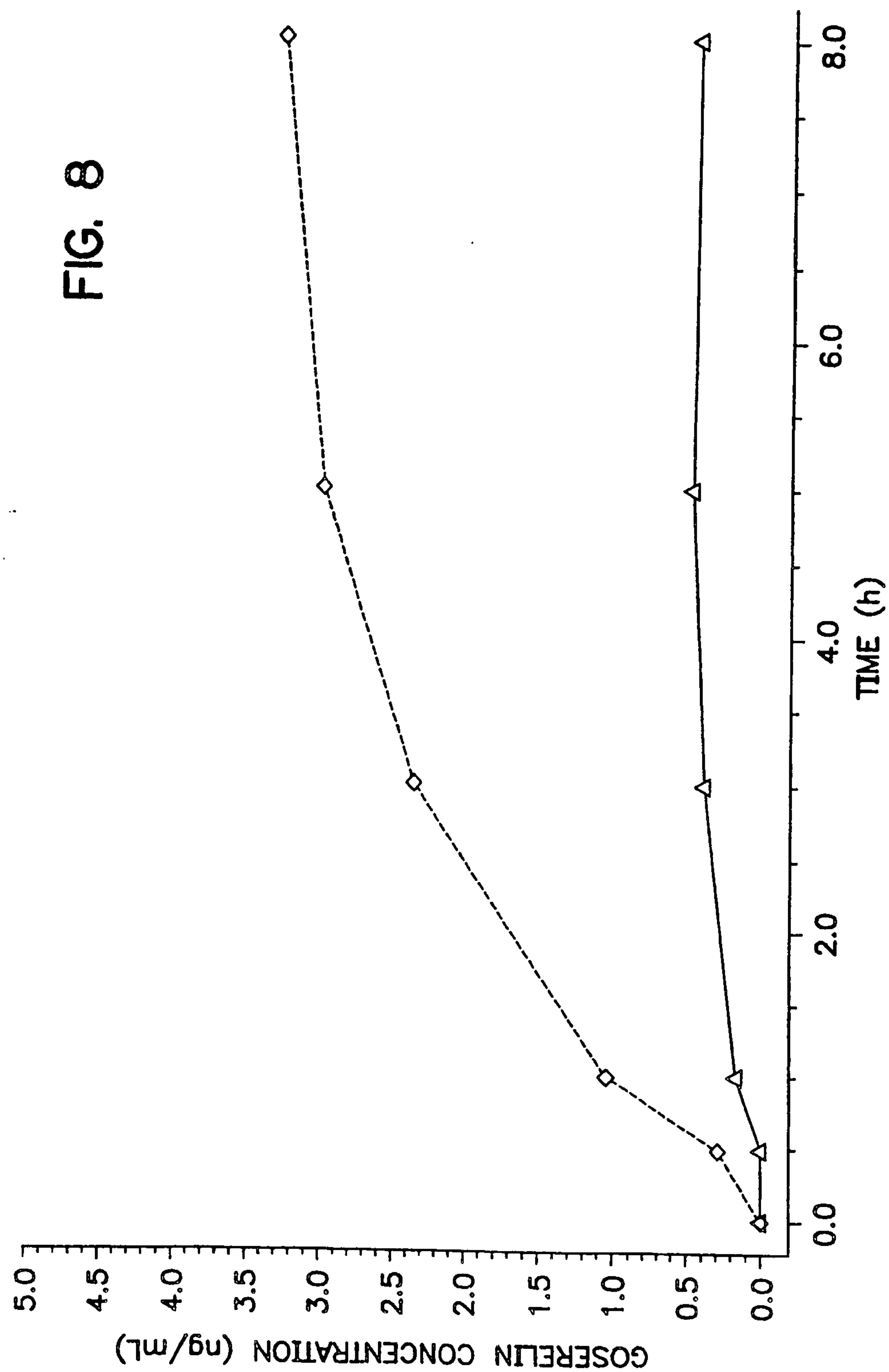
0.5
1.0
1.5
2.0
2.5
3.0

FIG. 8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0.5
1.0
1.5
2.0
2.5
3.0
3.5
4.0
4.5
5.0
TIME (h)
0.0
2.0
4.0
6.0
8.0

INTERNATIONAL SEARCH REPORT

International Application No
PCT/US 96/10128

A. CLASSIFICATION OF SUBJECT MATTER
IPC 6 A61N1/32

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 6 A61N

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	WO,A,91 15258 (MEDTRONIC INC) 17 October 1991 see page 4, line 18 - page 7, line 8; figures ---	1,2, 5-10,15, 19-23
A	WO,A,92 18197 (OPTISCHE IND DE OUDE DELFT NV) 29 October 1992 see page 3, line 14 - page 4, line 24; figures ---	1,5-11, 14,15, 19-24,27
A	EP,A,0 547 482 (BECTON DICKINSON CO) 23 June 1993 see page 5, line 18 - page 11, line 7; figures ---	1,5-10, 13-15, 19-23, 25-27

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[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

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"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed

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"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
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"&" document member of the same patent family

1

Date of the actual completion of the international search	Date of mailing of the international search report
12 November 1996	29. 11. 96

Name and mailing address of the ISA	Authorized officer
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INTERNATIONAL SEARCH REPORT

International Application No
PCT/US 96/10128

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

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A	PHARMACEUTICAL RESEARCH, vol. 8, no. 3, 1991, pages 365-369, XP002018300 M.J. PIKAL AND S. SHAH: "Study of the Mechanisms of Flux Enhancement Through Hairless Mouse Skin by Pulsed DC Iontophoresis" cited in the application -----	1,6-10, 14-16, 19-23,27

1

Form PCT/ISA/210 (continuation of second sheet) (July 1992)

INTERNATIONAL SEARCH REPORT

Information on patent family members

I ational Application No
PCT/US 96/10128

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		ES-T-	2072761	16-07-95
		JP-T-	6503496	21-04-94
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		CA-A-	2084734	18-06-93
		JP-A-	5245214	24-09-93
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		EP-A-	0313638	03-05-89
		JP-T-	2500339	08-02-90
		US-A-	4931046	05-06-90